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Basic Science

The effect of neighboring segments on the measurement of segmental stiffness in the intact lumbar spine

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Abstract

BACKGROUND CONTEXT: Degeneration, injury, and surgical interventions may alter the mechanical properties of spinal motion segments, but the quantification of these alterations in vivo is problematic. Manual or instrumented loading of single segments in the intact spine as applied intraoperatively may overestimate the mechanical properties of this segment, because the applied load is partly sustained by the adjacent segments.

PURPOSE: The distribution of stiffness values of individual spinal segments within and across spines was determined so as to use these data as input to a model simulation of segment stiffness tests in intact spines, to assess measurement errors.

STUDY DESIGN: Biomechanical stiffness measurements on human cadaveric spines and model simulation to assess measurement errors.

METHODS: Seventeen human cadaveric lumbar spines were loaded with pure moments in flexion/extension, lateral bending, and torsion. An optical system was used to measure the angular rotations of each motion segment and load-displacement curves were used to determine stiffness. With the distribution of measured stiffness data as input, a stochastic mechanical model was constructed to investigate how the stiffness of adjacent segments influences stiffness estimates obtained by loading a single segment in the intact spine.

RESULTS: The variance in stiffness values was high for all directions, but covaried between segments within a spine. Model simulations indicated that stiffness estimates obtained by loading a single segment in an intact spine are highly correlated with actual stiffness, but overestimate stiffness by a median of 18% with peak errors of close to 400%.

CONCLUSION: Current measurement devices and manual assessment substantially overestimate segmental stiffness due to the effect of adjacent spinal levels. In addition, the variance in stiffness within spines can occasionally cause large errors, which might lead to erroneous surgical decisions.

Keywords: Spine; Biomechanics; Intraoperative measurement; Adjacent segment; Stiffness; Diagnostics

Introduction

Degeneration, injury and surgical interventions, such as laminectomy, may reduce the stiffness of spinal motion segments [1,2], whereas reduced segmental stiffness has been suggested to prelude the occurrence of spinal deformities, such as degenerative scoliosis and spondylolisthesis [3–5], as well as low back pain [6]. Reduced stiffness of spinal segments is therefore implicitly or explicitly a factor in surgical decision making, for example, on whether or not to use instrumentation to stabilize a segment, or on which segments to fuse. However, the assessment of segmental stiffness in vivo remains problematic. Although magnetic resonance imaging (MRI) and radiographs can visualize degeneration, and kinetic MRI and flexion-extension

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radiographs can show altered motion in degenerated segments [6–8], they do not provide information on segmental mechanical properties.

Intraoperatively, the surgeon typically tests segmental stiffness manually by moving apart two neighboring vertebrae and estimating the stiffness from the tactile feedback. Efforts to make these assessments more objective and quantitative for research and future clinical applications have led to the development of intraoperative measurement devices. These devices typically consist of manually operated or motorized spinal spreaders and measure the force applied to the segment tested as well as its movement [7–15]. Segmental stiffness is then inferred from the relation between applied force and the resulting displacement.

Although the results of current measurement devices seem promising, they may not provide valid measurements of the mechanical properties of spinal motion segments. One limitation is that these devices apply a combination of an axial distraction force and a bending moment. This problem was elegantly solved by Reutlinger et al [15] through application of a method to separate the relative movement of the vertebrae due to axial translation and bending. A second limitation of these methods is that the motion segment under test is connected to adjacent segments, which will sustain a portion of the applied load. Not only the intervertebral joint tested, but also the adjacent intervertebral joint will deform and produce reaction forces and moments. Therefore, the measured force not only depends on the local segmental stiffness, but also on the stiffness of the neighboring segments. Reutlinger et al [15] acknowledged this limitation, but asserted that only a minor portion of the applied load will be borne by adjacent segments. To our knowledge, however, the magnitude of errors in the measurement of spinal stiffness due to load sharing by adjacent segments is at present unknown. Clearly, the magnitude of the load borne by adjacent segments will depend on the stiffness of these segments (see Fig. 1 for explanation). Therefore, we determined the stiffness of lumbar spinal motion segments in intact cadaveric human spines by applying pure moments, which avoids the problem mentioned previously, but can be done only in vitro and we analyzed the distribution of stiffness within and between spines. Subsequently, we used these data in a Monte Carlo simulation with a simple mechanical model of the spine to estimate the effect of the load borne by adjacent segments on the validity and precision of in vivo measurements of local spinal stiffness.

Materials and methods

Specimens

Seventeen fresh frozen human lumbar spines were used in this study (5× L2–L5 and 12× L1–L5). The spines were harvested from 12 male cadavers and 5 female cadavers with an age range of 55 to 90 years at time of death. The spines were wrapped in plastic bags and stored at −20°C.

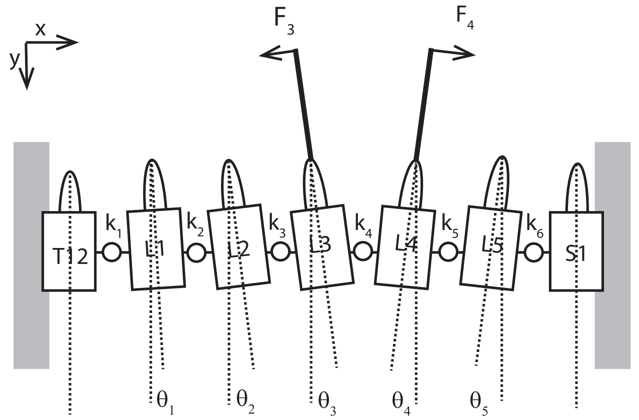

Fig. 1. Illustration of a single segment bending test in flexion direction at L3–L4. Forces are applied to the L3 and L4 vertebrae by rotating these in opposite directions. The rotations of the L3 and L4 vertebrae θ_3 and θ_4 cause rotations of the adjacent vertebrae as well. The rotation of these vertebrae is resisted by deformation and hence stiffness (k_4) of the L3–L4 intervertebral joint, but also by the deformation and stiffness (k_3 and k_5) of the L2–L3 and L4–L5 intervertebral joints.

Before testing, the spines were dissected and musculature was carefully removed, leaving the ligaments intact.

Mechanical testing

Eighteen hours before mechanical testing, the spines were thawed to room temperature and the upper and lower end vertebrae were embedded in cups fitting the testing machine using a low melting temperature alloy (Cerrolow-147: 48.0% bismuth, 25.6% lead, 12.0% tin, 9.6% cadmium, and 4.0% indium). Saline-soaked gauze was wrapped around the spines and sprayed with saline solution during preparation and testing to minimize dehydration.

A custom-made device (Fig. 2), previously described by Busscher et al [16], was used in which four-point bending could be applied for flexion-extension (FE) and right and left lateral bending (LB), as well as pure moments for right and left torsion (T). The device ensured that all segments experienced equal moments; therefore, differences in deformation of neighboring segments were determined by differences in mechanical properties only. The bending device was driven by a material testing system (Zwick Roell, Ulm, Germany, model TC-FR2.5TN, and 84 Instron & IST, model 8872, Norwood MA, USA). Before testing, an axial preload of 250 N was applied to the spines for 1 hour. Preload was then removed and three continuous load cycles from −4 Nm to +4 Nm were applied in each loading direction at a rate of 0.5 degrees per second. The order in which the loading directions were applied on the spines was balanced.

During mechanical testing, kinematic data were recorded with an opto-electronic system (Optotrak; Northern Digital, Waterloo, Ontario. Canada). Clusters of three infrared LED markers were rigidly fixed to the anterior surface of the vertebral bodies and were related to the anatomical

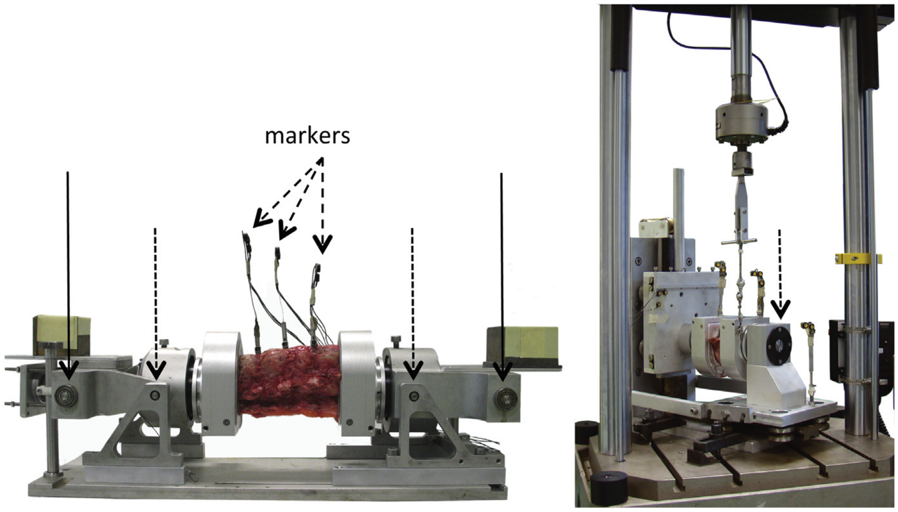

Fig. 2. (Left) Lumbar spine embedded in the test device for flexion/extension testing. Equal loads are applied at the points indicated by the solid arrows, while the device allows rotation around the axes indicated by the dashed arrows. For lateral bending, the specimens was rotated by 90°. (Right) The device placed in the testing machine for torsion testing. Load is applied through the cable, while the device allows rotation around the axis indicated by the dashed arrow.

axes of motion of the spines. This was done by making a series of short recordings, while placing a probe containing six markers at anatomical landmarks. This allowed relating the positions of the clusters to the anatomical landmarks and hence the estimated rotation axes.

Data analysis

Kinematic data of L2–L3, L3–L4, and L4–L5 were extracted using a computer program written in Matlab (Mathworks, Natick MA, USA). The mechanical properties were assessed using the method of Smit et al. [17], which makes use of a double sigmoid function that is fitted over the raw load-deflection data to filter noise and to allow for an analytical calculation of the nonlinear segment compliance. The region with the highest compliance represents the neutral zone and its boundaries were determined by assessment of the maximum and minimum of the second derivative. The neutral zone stiffness (k_{NZ}) was calculated as the slope of the neutral zone (Fig. 3). In this study, only the trials for which the correlation between measurement data and the fitted curve was 0.90 or greater were considered eligible.

Statistical analysis and simulation of bending tests

The statistical analysis of the stiffness values obtained was aimed at characterizing the distribution of these values as a basis for the simulation. Potential sources of variance considered were spinal level, spine (ie, random between-subject variance), and random variance within spines. First, the distribution of stiffness was inspected. Given the skewness of the distribution, analysis of variance was performed for each loading direction on logarithmically transformed stiffness values, with spine as a random factor and spinal level (L2–L3, L3–L4, and L4–L5) as a fixed factor, for each loading direction separately. Subsequently, means and standard deviations (SDs) of the stiffness values per spine were determined and were plotted against each other to assess the presence of heteroscedasticity (ie, dependence of the SD on the mean).

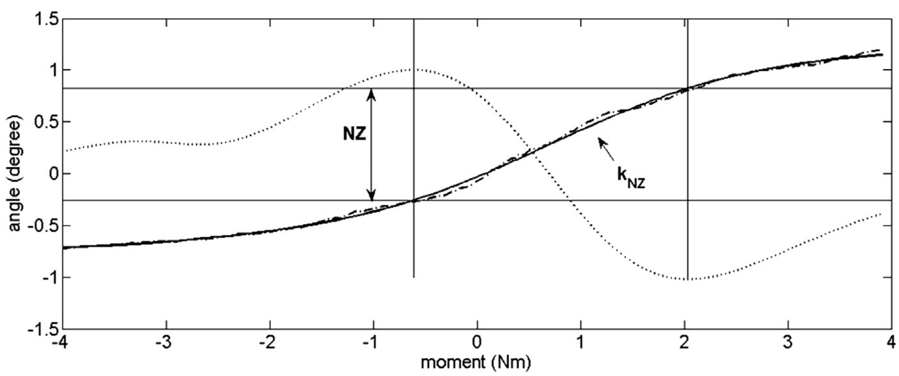

Fig. 3. Typical load-deflection curve from a quasi-static bending test. The dash-dotted line represents the actual data, the solid line is the curve fitted to the data, and the dotted line is the second derivative of the fitted curve. The neutral zone (NZ) is based on the maximum and minimum in the second derivative, and neutral zone stiffness (k_{NZ}) is calculated as the slope of the NZ.

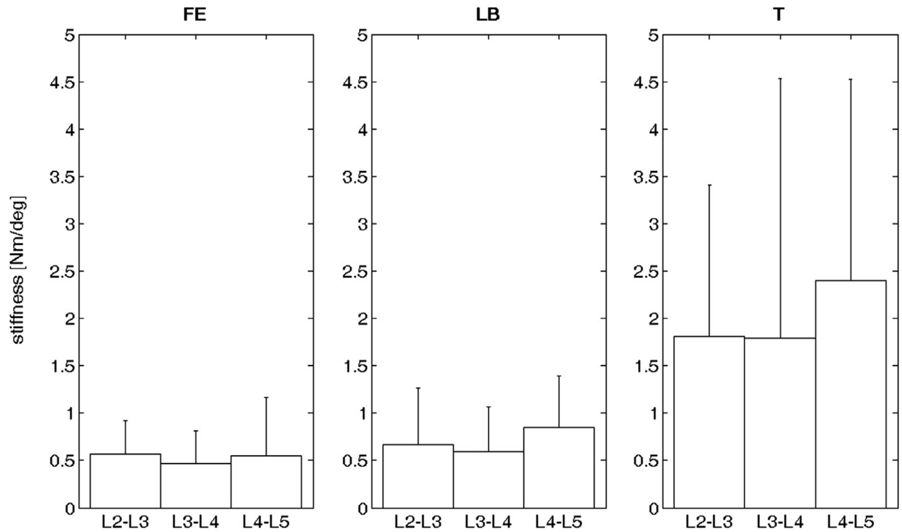

Fig. 4. Neutral zone stiffness (k_{NZ}) of L2–L3, L3–L4, and L4–L5 segments in flexion-extension (FE), lateral bending (LB), and torsion (T). Error bars indicate standard deviations. The differences in height between the bars indicate a difference in kinematic behavior between anatomical levels, although these differences were not statistically significant.

The effect of stiffness variations along the spine on the stiffness estimates obtained by single-segment testing was assessed with a biomechanical model. The lumbar spine was modeled as a series of rigid bodies representing vertebrae T12 to S1, linked by rotational springs, representing the intervertebral joints (for details see Appendix 1). A statistical simulation procedure was used to simulate a large number of spines based on the statistical properties of the measured data (for details, see Appendix 1). In short, 1,000 samples were drawn (with replacement) from the means within spines. Each mean stiffness drawn defined the mean stiffness of the simulated spine, whereas the stiffness of the individual segments was randomly chosen based on this mean and the concomitant SD. For each simulation, joint angles resulting from a 1-Nm load applied at a specific lumbar level were calculated. Subsequently, the stiffness was estimated from the deformation at that level (ie, ignoring the effect of adjacent segments) and the result was compared with the stiffness value used as model input. The procedure was repeated for each of the six lumbar intervertebral joints and each movement direction. As outcome variables, we determined for each simulation the (relative) estimation error and the Spearman rank order correlation between estimated stiffness values and input stiffness values.

Results

Motion segment stiffness

The values for k_{NZ} for four segments in FE, five segments in LB, and four segments in T were excluded, because correlations between measured data and curve fitted data were less than 0.90. The mean values for k_{NZ} for all segment levels and all directions are shown in Fig. 4.

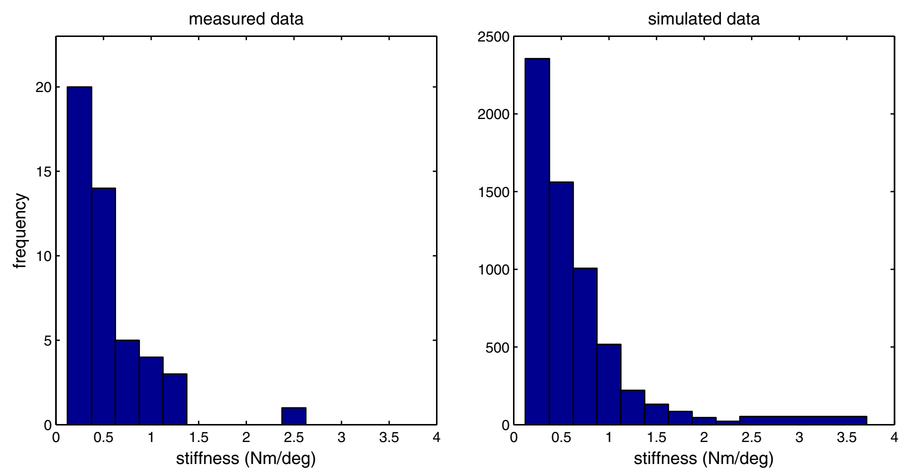

Fig. 5. Histogram of all measured flexion-extension (FE) stiffness values (Left) and all FE stiffness values used in the simulations (Right) based on the simulation procedure.

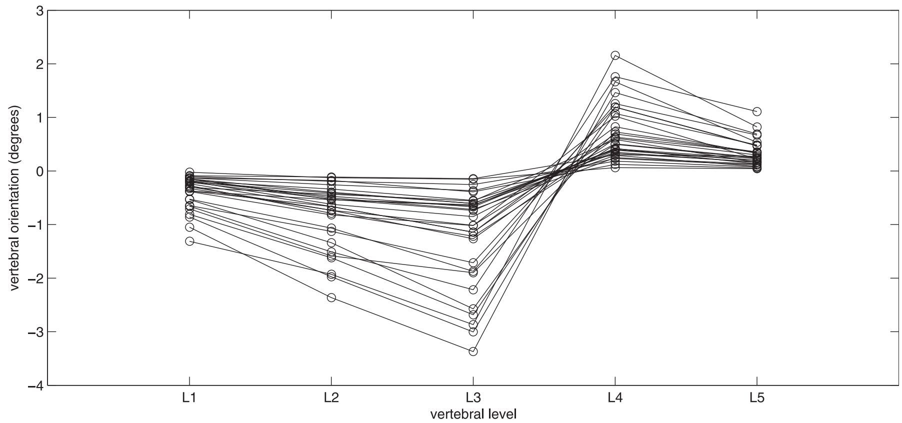

Fig. 6. Rotation in degrees per vertebra in 30 simulations of a single-segment flexion-extension bending test at L3–L4.

Segment level had no significant effect on stiffness in any movement direction (p>.23). On the other hand, differences between spines were highly significant. The SD of the stiffness values within spines was significantly correlated to the mean stiffness for all movement directions (r=0.84, 0.86, and 0.74 for FE, LB, and T, respectively, all p≤.001). Coefficients of variation were on average 33% (SD 15%), 53% (SD 30%), and 48% (SD 30%), for FE, LB, and T, respectively.

Simulation of single-segment tests

Given the lack of effects of segment level, the simulations assumed a random distribution of stiffness within spines, while in view of the heteroscedasticity of the data, concomitant values of the mean and SD were used to define the distribution of stiffness across segments within spines. The simulation procedure resulted in a distribution of stiffness values that closely resembled the distribution in the measured data (Fig. 5), justifying these assumptions on distribution and the mixed bootstrap and Monte Carlo procedure.

Results of the simulations were comparable across segment levels and movement directions and therefore only the results for FE of the L3–L4 segment are presented in detail. Fig. 6 illustrates the bending pattern of a series of simulated spines when loading the L3–L4 segment in FE. As might be expected when the L3–L4 segment is bent, the cranial vertebrae also move with the L3 vertebra (negative angles), whereas the caudal vertebrae move with the L4 vertebra (positive angles) across all simulations. Differences between the simulations are seen in the degree of motion across the segments and depend on the stiffness values used as input.

Simulated single-segment tests resulted in a consistent overestimation of the true stiffness, whereas the correlation between actual and estimated stiffness was high (all r>0.99; Fig. 7, Table). The estimation error was dependent on the actual stiffness (Fig. 6) and consequently both absolute and relative estimation errors showed a skewed distribution with a wide range (Fig. 8, Table). The high correlation between actual and estimated stiffness led to perfect identification of the rank

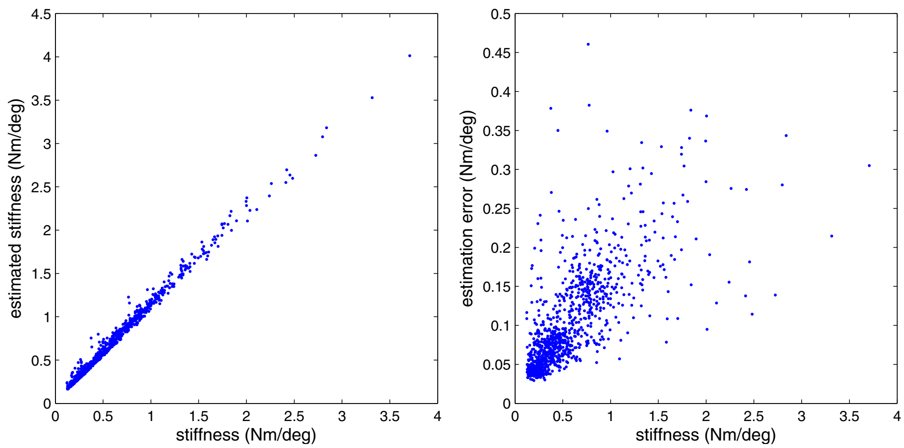

Fig. 7. Estimated stiffness in simulations of single-segment flexion-extension bending tests at L3–L4 as a function of the actual stiffness (Left). Estimation errors (ie, difference between estimated and actual stiffness) as a function of the actual stiffness (Right).

Table
Median (range) of absolute and relative estimation errors (ie, difference between estimated and actual stiffness) obtained in simulated single-segment tests in an intact spine for each and movement direction pooled over segment levels

Movement direction	Error, Nm/deg	Relative error, %
Flexion extension	0.079 (0.031–0.410)	18.7 (2.8–147.5)
Lateral bending	0.122 (0.031–1.058)	17.9 (1.4–377.9)
Torsion	0.301 (0.037–2.220)	18.6 (2.4–268.2)

order of stiffness values across segment levels within spines based on single-segment testing in all simulations.

Discussion

The assessment of the presence, location, and severity of segmental instability is assumed to be relevant in planning interventions and evaluating treatment. Unfortunately, the assessment of segmental stiffness in the intact spine is problematic. This study was performed to determine to what extent such measurements of segmental stiffness are affected by load sharing by the neighboring segments. The results showed that tests on single segments in an intact spine yield a systematic overestimation of stiffness with a median of 18%, but with occasional very large overestimations (>300%). The systematic overestimation is a consequence of the fact that adjacent segments resist part of the applied load in single-segment testing in the intact spine. Large random errors occur, because in some cases a segment with low stiffness is surrounded by much stiffer segments, in spite of the correlation between segmental stiffness values in spines.

Although stiffness values were correlated within spines, the measurements revealed a substantial heterogeneity of kinematic behavior between segments under pure moment loading of the lumbar spine, implying that within-spine differences in k_{NZ} are present between segments. Previously, Panjabi et al [18] reported a significant influence of spinal level on the mechanical properties; specifically, they reported that in flexion the motion of L4–S1 was greater than of L1–L3, and in T the motion of L2–L3 was greater than of L4–S1. These results were not reproduced by this study, which is inconsistent with the notion that every spine shows a gradual cranial-caudal variation in anatomical and mechanical properties [19]. However, a gradual variation might be found only in healthy young spines. When aging or degeneration of spinal elements does not start at the same time at all anatomical levels, or does not advance at the same rate, this systematic gradual variation might be altered.

Recently, a number of different measurement devices have been developed with the aim of measuring segmental stiffness at a specific level [7–13,15,20–22]. The present study showed that these methods can yield substantial overestimates, especially when adjacent segments have a high stiffness. Hasegawa et al [12] performed measurements in patients with spinal pathology and reported a high stiffness for asymptomatic segments adjacent to previously operated segments. However, because they did not consider the influence of neighboring segments, it is not clear whether the asymptomatic segments actually were stiffer or that their stiffness was overestimated, due to the presence of an adjacent much stiffer (fused) segment. In our view, the measurement errors predicted by our model are not acceptable for use in a scientific context. Although one might argue that tests on single segments in the intact spine may be useful in clinical practice, given the high correlation between actual and estimated stiffness. This would be true when decision making is based on a comparison of several segments within a spine or on measurements of the same segment before and after an intervention. However, when results are compared to an external standard, such as the surgeon's previous experience, occasionally large overestimation of stiffness may occur, when much stiffer segments surround the segment tested. This might lead to erroneous decisions; for example, not to use

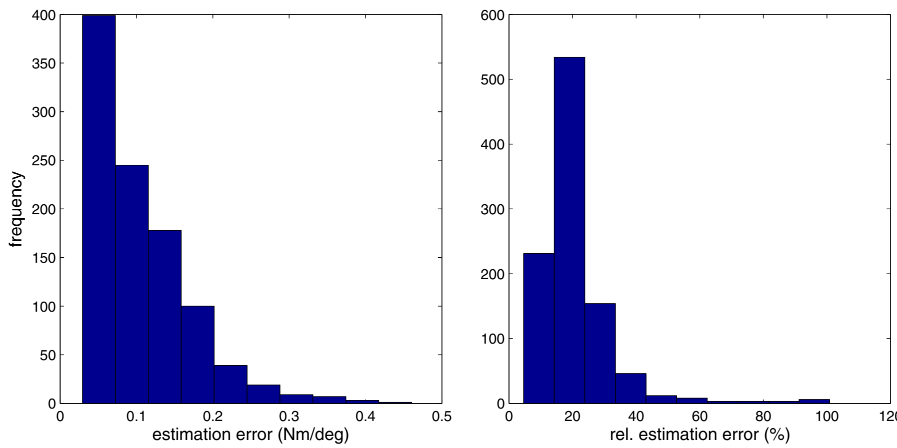

Fig. 8. Histogram of absolute (Left) and relative (Right) errors in stiffness estimates (ie, difference between estimated and actual stiffness) obtained in simulations of single-segment flexion-extension bending tests at L3–L4. rel., relative.

instrumentation to stabilize a segment after laminectomy, because its stiffness appears sufficient, due to load sharing by adjacent segments, although actually it is quite low.

The model that we used to point out possible errors in stiffness estimation is a simplified representation of the complex mechanics of spinal motion. First, the mechanical properties of the intervertebral joints were linearized. Linearization is allowed within the neutral zone, but outside the neutral zone, a clear nonlinearity can be observed (Fig. 3). Although an overestimation of stiffness is expected outside of the neutral zone as well, results may be different quantitatively. Furthermore, the model allows rotation only in the loaded direction. In reality, some translation will occur, which contributes to estimation errors in current testing methods when not accounted for [15]. Finally, in the model, T12 and S1 were the rigidly fixed boundaries of the system, whereas in reality these segments are mobile and connected to adjacent structures with a specific stiffness. Still, this might not have had major effects on the results, as rotations of vertebrae more distant from the segment tested are limited (Fig. 6).

Conclusion

The results of this study show that neighboring spinal motion segments within the same spine can have a substantially different stiffness. Model simulations indicate that load sharing by adjacent segments causes an overestimation of segmental stiffness with a median of 18% when assessed with current intraoperative measurement devices and manual testing. When adjacent stiffness values are high, errors of more than 300% may occur. Therefore, further research efforts should be directed toward the development of alternative approaches toward measuring the mechanical properties of individual motion segments in vivo.

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Appendix 1

The lumbar spine was modeled as a series of rigid bodies (vertebrae T12 to S1), linked by rotational springs, representing the intervertebral joints ($k_1 \ldots k_6$). The T12 and S1 vertebrae were assumed to be rigidly fixed (Fig. 1). A spreader was assumed to exert equal but opposite forces (F=25 N) on two adjacent vertebrae (L3 and L4) over moment arms that were considered to be the same for both vertebrae (a=0.04 m), resulting in a moment of 1 Nm. This moment causes rotation of vertebrae L1 to L5 ($\theta_1 \ldots \theta_5$). The relation between force, lever arm, displacement, and stiffness is given by Equation (1). Note that the same equation was used to model lateral bending and torsion with stiffness ($k_1 \ldots k_6$) set at appropriate values for these directions.

$$\begin{bmatrix} 0 \\ 0 \\ aF_3 \\ aF_4 \\ 0 \end{bmatrix} = \begin{bmatrix} k_1+k_2 & -k_2 & 0 & 0 & 0 \\ -k_2 & k_2+k_3 & -k_3 & 0 & 0 \\ 0 & -k_3 & k_3+k_4 & -k_4 & 0 \\ 0 & 0 & -k_4 & k_4+k_5 & -k_5 \\ 0 & 0 & 0 & -k_5 & k_5+k_6 \end{bmatrix} \begin{bmatrix} \theta_1 \\ \theta_2 \\ \theta_3 \\ \theta_4 \\ \theta_5 \end{bmatrix} \tag{1}$$

Equation (1) expresses the dependence of the local deformation on the stiffness of the local level as well as on the stiffness of the adjacent levels. For example, the relation between a force applied to L4 and the rotation of L4, derived from Equation (1) is:

$$aF_4 = -k_4 \bullet \theta_3 + (k_4 + k_5) \bullet \theta_4 - k_5\theta_5, \tag{2}$$

which shows that k_4, the stiffness of the L3–L4 segment, cannot be determined without measuring rotations θ_3, θ_4, and θ_5 (ie, rotations of L3, L4, and L5) and without estimating k_5. When the load-deformation data from a test on a single segment *would* be used to estimate its stiffness, this would result in the incorrect estimate k'_4:

$$(aF_3 + aF_4)/2 = k'_4(\theta_4 - \theta_3) \tag{3}$$

In this estimate, the fact that the rotations θ_3 and θ_4 of L3 and L4 also depend on the stiffness of L2–L3 and of L4–L5 (k_3 and k_5) is not taken into account.

The effect of stiffness variations along the spine on the stiffness estimates obtained by single-segment testing was assessed with the model. Stiffness values $k_1 \ldots k_6$ were assigned to each of the joints by means of a mixed bootstrap [23] and Monte Carlo simulation. Because of the skewness of the distribution of stiffness values across spines, means and concomitant standard deviations (SDs) per spine were subjected to a nonparametric bootstrap analysis. Samples were drawn from the original data (pairs of means and SDs of stiffness within a spine) with replacement. Each mean drawn defined the center of a normal distribution, with a spread around that center based on the concomitant SD. Subsequently, a random sample of six stiffness values was generated based on this distribution (Monte Carlo simulation). Samples containing a stiffness value lower than the minimum stiffness measured (over all segments), were discarded. For each sample, joint angles resulting from a 1-Nm load applied at a specific lumbar level were calculated. Equation 1 or modifications thereof to simulate load application at other levels than L3–L4 (hence changing the left hand matrix only) were used to determine resulting rotations. Using Equation 3 with these rotations as input, a stiffness estimate for the loaded segment was obtained ($k_1 \ldots k'_6$) and compared with the input values ($k_1 \ldots k_6$). This procedure was repeated 1,000 times for each of the six lumbar intervertebral joints to determine accuracy and precision of $k'_1 \ldots k'_6$.